Fluid-structure interaction analysis of hyperelastic shells subjected to blood flow

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ABSTRACT

Aortic dissection (AD) is a life-threatening medical condition which occurs when a tear in the intima, the inner layer of the aortic wall results in blood escaping and an intramural hematoma develops [1]. In addition to clinical evaluations, the medical profession have at their disposal a series of computational tools that can help in assessing the severity of the patient condition. However, running and analyzing the results is a vastly complex task that requires special training [2].

In this work, we use a three-dimensional fluid-structure interaction (FSI) analysis to gain better insight into the mechanical behavior of a healthy human aortic segment when subjected to blood flow. Hyperelastic constitutive equations are known to better represent biological tissues [3, 4], and are used to model the human aortic segment in a cylindrical shell form. The walls are represented by three layers with material properties that vary drastically. The fluid flow is modelled as pulsatile, Newtonian, incompressible and turbulent to mimic realistic physiological conditions and the fluid domain is also represented by a cylindrical domain. The material properties are summarized in Table 1.

Figure 1 depicts the discretized cylindrical shell used for the 3D-FSI simulations. Figure 2 shows the radial displacement obtained with the hyperelastic model for the solid structure and subjected to fluid flow as described above. Results are compared with those obtained previously with an elastic model. The profile are similar for both cases with a maximum amplitude at half-period. However, it is seen that the elastic model underestimates the radial displacement by about 40 % (1.7 mm for elastic vs. 2.4 mm for hyperelastic). For the stress-strain behavior (Fig. 3), there is a sharp contrast between the two models: the hyperelastic case exhibits a clearly non-linear behavior and a large change of strain for a small stress variation. When elastic constitutive equations are taken to model the cylindrical shell, the stress-strain behavior is linear.

In terms of computational efforts, it takes approximately 5 hours to run two cardiac cycles with the elastic model and 8 hours with the hyperelastic model (for the same number of elements). Additional work is under progress to evaluate the trade-off between computational time and the potential gain in using an hyperelastic model.

Table 1: Geometry and hyperelastic material properties of human aorta [3]

	Material	Thickness t (mm)	Fiber orientation θ (deg)	μ (kPa)	k_1 (kPa)	k_2	Poisson's ratio
Human Aorta (3 layer, hyperelastic, anisotropic cylindrical shell)	Intima	0.33	18.8 ± 8.2	20.30	182.59	228.58	0.49
	Media	1.32	37.8 ± 20.6	2.31	8.45	12.84	0.49
	Adventitia	0.96	58.9 ± 14.8	6.16	32.89	167.31	0.49

Where μ, k_1 and k_2 are shear modulus, stress-like parameter and dimensionless parameter respectively.

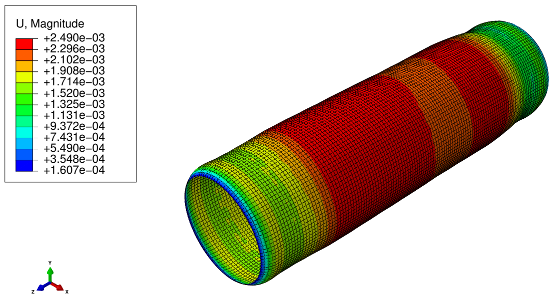

Figure 1: Discretized model of the human aortic segment used for the 3D-FSI simulations.

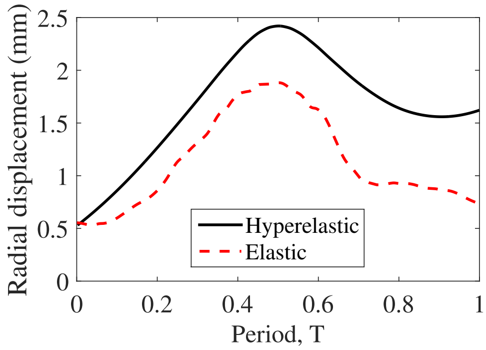

Figure 2: Radial displacement of healthy human aortic segment subjected to blood flow.

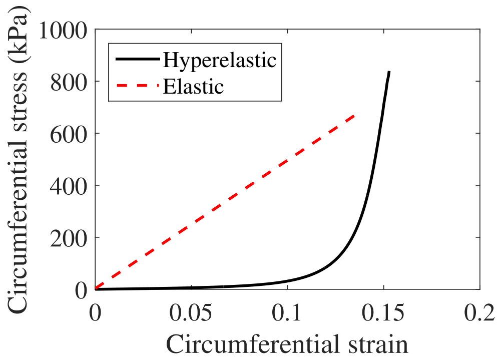

Figure 3: Hoop stress-strain profile obtained with elastic and hyperelastic model.

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